The Predictive Value of Stress Shielding for Quantification of Adaptive Bone Resorption Around Hip Replacements

The presence of a femoral hip stem changes local mechanical signals inside the surrounding bone. In this study we examined the hypothesis that the eventual loss of bone can be estimated from the initial patterns of elastic energy deviation, as determined in FE models of the intact bone and the operated femur. For that purpose two hypothetical relations between elastic energy reduction and resorption were investigated. Their estimates of bone loss were compared to the results of iterative computer simulations. Two kinds of FE model were used, and in each stem stiffness and remodeling threshold (a measure of "biological reactivity") were varied. Provided that reasonable values of the remodeling threshold are assumed and that the stem is firmly bonded to the bone, we found that the difference between direct estimates and simulation models was 4 percent of bone loss. It is therefore concluded that initial patterns of elastic energy deviation give a reasonable indication of expected bone loss.

Methods

We assumed that all bone will be resorbed when the mechanical signal 5 after insertion of the prosthesis is zero. Conversely, all bone will be preserved when the signal is equal to that prior to insertion of the prosthesis, the "natural" or reference signal 5n.f. Around the reference signal a threshold level s was assumed (Carter, 1982, 1984; Frost, 1983, 1987; Huiskes et al., 1987, 1992; Nauenberg et al., 1993). Mechanical signal reductions that do not reach the threshold level will not cause bone resorption. Using an "influence function" g(S), which has a value of one when the signal is zero and a value of (almost) zero when the signal is equal to the reference signal minus the threshold, the total resorbed bone mass M was obtained by integrating the product of original bone density p and influence function g over the total bone volume Ω. The resorbed bone mass fraction m was obtained by normalizing M, to the original bone mass M:

\[ m_r = \frac{1}{M} \int \rho(S(x)) g(S(x)) dx \]  \hspace{1cm} (1)

where:

- \( m_r \) = resorbed bone mass fraction
- \( M \) = original bone mass
- \( \Omega \) = original bone volume
- \( S(x) \) = local value of mechanical signal
- \( \rho(x) \) = local apparent bone density

Two influence functions were tested. The first function is the normal or Gaussian cumulative distribution function (Fig. 1, curve A):

\[ g_1(S) = 1 - \frac{1}{\sigma \sqrt{2\pi}} \int_{-\infty}^{-\frac{S - S_{ref}}{\sigma}} \exp\left[-\frac{1}{2}\left(\xi - \mu\right)^2\right] d\xi \] \hspace{1cm} (2)

where:

- \( \mu \) (mean) = 0.5(1.2 - s)S_{ref}
- \( \sigma \) (standard deviation) = 0.1μ

The second influence function is a step function (Fig. 1, curve B):

\[ g_2(S) = \begin{cases} 
1 & \text{for } \frac{S - S_{ref}}{S_{ref}} < -s \\
0 & \text{for } \frac{S - S_{ref}}{S_{ref}} > -s 
\end{cases} \] \hspace{1cm} (3)

Following Carter et al. (1989), Huiskes et al. (1989, 1992),...
Two FE models were used. The first was two dimensional and represented a 'straight stem' loaded with a bending moment of 1000 Nmm (Huiskes et al., 1987, Fig. 2(a)). The second model was three dimensional. The geometry and density distribution from the bone were determined from CT scans (Huiskes et al., 1992; Fig. 2(b)), and a cycle of three hip joint and muscle loading cases was used representing a daily loading pattern (Carter et al., 1989). For both models, a fully bonded 'iso-elastic' (E = 20,000 MPa) and titanium (E = 115,000 MPa) prosthesis was modeled. For the two-dimensional model, the normalized amount of stress shielding (S = Sref)/Sref was far from uniform (Fig. 4). Three remodeling simulations were performed, two using a titanium stem (s = 0.35 and s = 0.75) and one using an iso-elastic stem (s = 0.75; Huiskes et al., 1992). The Gaussian function gave close agreement between estimation and simulation in all regions, regardless of threshold level or prosthetic stiffness. The maximum deviation was 2 percent for total bone loss (iso-elastic stem), 8 percent at level 1 (iso-elastic stem), 12 percent at level 2 (titanium stem, s = 0.75), 4 percent at level 3 (titanium stem, s = 0.35 and s = 0.75) and 5 percent at level 4 (titanium stem, s = 0.35). With the step function, estimation and simulation only came close for s = 0.75. The maximum deviations occurred for the titanium stem and were 4 percent for total bone loss, 6
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The step curve gave only close approximations of bone loss for
the larger threshold level of 75 percent. Apparently, such a large
threshold zone is required for a realistic
simulation of the remodeling process. The value of the threshold
level is not known. Engh and Bobyn (1988) estimated that a
loss of 30–45 percent of the bone stresses (or 75 percent of
the strain energy density) seems tolerable, which would suggest
a level of 0.7–0.8. Huiskes et al. (1992) found that a threshold
zone of 0.75 was required to predict realistic resorption patterns.

Discussion
Bone remodeling is a nonlinear process, during which altered
mechanical signals change bone density patterns, leading in
their turn to an altered local mechanical signal. Phenomena may
therefore occur later that cannot be deduced from the initial
situation, in particular around unbonded prostheses. But even
when the prosthesis is bonded, resorption of bone in one location
imposes higher loads on bone elsewhere, which may subsequently
reduce the amount of bone lost. The step function (Eq. (3))
completely neglects this kind of nonlinearity. The Gaussian
distribution curve (Eq. (2)) partially accounts for it, but bone
apposition is neglected.

Nonlinearity depends mainly on the nonuniformity of the
initial signal distribution, in association with the threshold level
$s$. The simple two-dimensional configuration hardly showed
any nonlinearity, and the estimate of bone loss using the step
function came closest to the remodeling simulation. In this case
the Gaussian curve did not perform well at all. Conversely, for
the three-dimensional configuration the Gaussian curve gave a
close approximation to the remodeling simulation in all cases.
The step curve gave only close approximations of bone loss for
the larger threshold level of 75 percent. Apparently, such a large
threshold reduces nonlinearities to a sufficiently low level.

The notion of a threshold was first introduced by Carter
from observations by Maloney et al. (1989), Engh and Bobyn
(1988), and Nauenberg et al. (1993). Theoretical support came
from Huiskes et al. (1987, 1992) and Weinans et al. (1993),
who showed that a threshold zone is required for a realistic

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